at a large number of sites, even though the scatterer concentration is low. These scattering sites are clustered in certain areas which depend on the scatterer size and arrangement. The degree of the clustering, not the number of scattering sites, obviously will be a good parameter for estimating scatterer concentration. When the number of scattering sites within a resolution cell has a probability given by a negative binomial distribution, we may use a parameter, α, to characterize the clustering. The parameter α can be found from higher order moments of echo amplitudes. From phantom experimental results, it is shown that the parameter, α, from the model is linearly proportional to the log scaled scatterer concentration in a range from 2 to 60 scatterer/mm^3, when the diffraction effect has been controlled.

7.4 QUANTITATIVE CHANGES IN ULTRASONIC BACKSCATTER FROM HUMAN SKELETAL MUSCLE WITH CONTRACTION, Maria Helguera and Jack G. Mottley, Department of Electrical Engineering, College of Engineering and Applied Sciences University of Rochester, Rochester, NY 14627.

Preliminary studies have shown that variations in ultrasonic properties such as backscatter and attenuation of skeletal muscle may be related to the physiologic state of those tissues, therefore holding great promise as indices for quantitative tissue characterization. These parameters may provide clinicians with valuable information about the condition of those tissues and eventually be of use in the diagnosis and monitoring of muscle diseases without the need for biopsies. This study was designed to elucidate patterns of change with contraction in ultrasonic backscatter from skeletal muscle in normal volunteers. Subjects were seated with their arm at 30°abduction and 90° flexion and asked to relax and then contract their bicep with the forearm held against a fixed object. A water-filled standoff with the ultrasonic transducer (10 cm focal length, broadband, 5 MHz center frequency) was applied to the anterior of the upper arm over the belly of the biceps, perpendicular to the skin surface. The backscatter spectrum from a 3.75 mm portion of muscle 1.51 $\pm$ .27 cm (mean $\pm$ SD) deep was recorded from five cycles of relaxation/contraction at each of nine sites. (Freq. range: 2-8 MHz in 0.25 MHz steps). The spectra were reduced to decibels relative to a planar, stainless steel reflector and the frequency averaged, or integrated, backscatter was calculated. Average integrated backscatter for relaxed and contracted muscle was obtained by averaging over all relax/contract cycles at each site, then all sites within a subject, then over all subjects. The values obtained for relaxation and contraction were -64.6 $\pm$ 1.16 dB and -67.2 $\pm$ 1.47 dB (mean $\pm$ sem, n = 5) respectively. The average change of backscatter with contraction over all volunteers was -2.57 $\pm$ 0.44 dB (mean $\pm$ sem, n = 5, $p < .05$). Discernible changes of backscatter with contraction are present in normal skeletal muscle and, if proven to correlate with pathological state, may prove useful in the diagnosis and monitoring of muscle diseases.

This work was supported in part by a grant from the Whitaker Foundation.

WEDNESDAY, JUNE 13

8. IMAGING

8.1 A COMPUTATIONAL AND EXPERIMENTAL STUDY OF NONDIFFRACTING TRANSDUCER FOR MEDICAL ULTRASOUND, J-y Lu and J.F. Greenleaf, Biodynamics Research Unit, Department of Physiology and Biophysics, Mayo Clinic/Foundation, Rochester, MN 55905.

Historically, transducers having Gaussian shading and Fresnel shaped voltage drives on an array of annular elements have been used for medical imaging of the highest quality. These transducers, have good focus within the depth of field and smooth near field patterns. Outside of the depth of field, the focus degrades with diffraction effects. Nondiffracting solutions to the wave equation governing propagation in tissues (the scalar Helmholtz equation) have recently been discovered and extensively tested on electromagnetic waves.

We have designed and fabricated a nondiffracting transducer for medical ultrasound. It was developed from PZT ceramic/polymer composite material and consisted of ten annular rings. The center frequency of the transducer was 2.5 MHz, and its bandwidth was around 50% of center frequency. Computer simulations and experimental measurements for the field of the nondiffracting transducer were carried out. Both continuous and pulsed wave results were obtained and compared to conventional Gaussian shaded beams. The nondiffracting beam has about 1.27 mm radius with a 20 cm depth of field compared to the Gaussian transducer of the same size with a 1.27 mm radius main lobe at a focus of 12 cm and 2.4 cm depth of field. The side lobes the nondiffracting beam are the same as the J_0 Bessel function. Phase aberration of the beam due to biological tissue in the nondiffracting transducer was found to be similar to that of the Gaussian beam. The new beam forming method may have promise in novel imaging and tissue characterization modes whereby the effects of diffraction, and the need for dynamic focusing are eliminated.

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8.2 PHASED ARRAY IMAGING WITH A COPOLYMER TRANSDUCER, S.W. Smith,[1,2] R.L. Goldberg,[3] L.F. Brown,[4] J.B. Castellucci,[3] J.J. Paulos[5] and O.T. von Ramm,[3] [1]Center for Devices and Radiological Health, Food and Drug Adminstration, Rockville, MD 20857, Departments of [2]Radiology and [3]Biomedical Engineering, Duke University Medical Center, Durham, NC 27706, [4]Atochem N.A. (formerly Pennwalt Corp.), Piezo Film Sensor Division, Valley Forge, PA 19482 and [5]Department of Electrical and Computer Engineering, North Carolina State University, Raleigh, NC 27695.

Pulse-echo phased array images have been obtained using a 32 element P(VDF-TrFe) copolymer transducer. The array was fabricated using 25 μm Pennwalt copolymer film with an aluminum backing and is quarter wavelength resonant at 24 MHz. Array elements were created by laser ablation of the continuous gold electrode. Thirty-two narrowband transmitters drive the transducer at 3.1 MHz. Custom I.C. buffer amplifiers process the received signal from 16 array elements. The amplifiers use bipolar junction transistors and have approximately unity gain. The amplifiers, which are mounted to the array connector, have a high input impedance compared to that of the array elements, 18 pF measured at 1 kHz. The amplifier output impedance matches the impedance of the coax cable, 75 ohms. The performance of the copolymer array was compared to that of a ceramic PZT array resonant at 3 MHz. The sensitivities of the two materials were studied with a hydrophone probe which was used to transmit 3 MHz bursts. Each receive channel consisted of a single array element followed by the buffer amplifier. In receive mode only, the copolymer sensitivity was 16 dB less than that of the PZT. This was caused partially by the low capacitance of each element and partially by driving the copolymer off-resonance. Each transducer was used to obtain pulse-echo phased-array images of a

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